# Method of Magnetic Resonance Imaging

The invention relates to improvements in and relating to magnetic resonance imaging (MRI), in particular to generation of magnetic resonance images of invasive devices, e.g. during surgical procedures.

During surgical and therapeutic procedures it is frequently desirable for the physician to be able to locate or guide invasive devices inserted into the body (for example catheters, guide wires, biopsy needles, etc.) when these are not directly visible to the naked eye.

Due to the reduced invasiveness of surgical procedures, MRI-guided interventional procedures have gained increasing importance in recent years. Such MRI guided procedures can be divided into two categories: intraoperative procedures which integrate surgery with MRI and interventional procedures for guiding, monitoring and controlling therapy.

Intraoperative procedures generally require an open magnet MR imager and are desirable as the MRI can be used to define anatomy and monitor tissue function as it changes during surgery. By monitoring anatomy and function in this way, the clinical outcome for the patient may be improved since complications may be reduced by reducing the degree of invasiveness of the surgery.

Interventional procedures generally require only limited patient access and thus conventional closed magnet MR imagers may be used. The high spatial and temporal resolution available in MRI allows for accurate "near real time" guidance of devices such as catheters, guide wires, biopsy needles, etc.

The success of an MRI-guided interventional or intraoperative procedure generally depends on the ability of the MRI technique to provide sufficiently accurate visualisation of the instruments and devices

inserted into the patient's body. Currently, either
active or passive visualisation techniques have been
used for monitoring and visualising such instruments and
devices (herein generally referred to as "invasive
devices").

In active visualisation, the invasive device is
provided with a small signal receiving antenna (e.g. on
the tip of the device). The magnetic resonance signal
from water protons in the vicinity of the antenna is
detected by the antenna and incorporated into the MR
image generated by the MR imager. The result is a "road
map" showing the movement of the antenna (and hence the
device) within the patient. There are however two
problems associated with active visualisation. Firstly,
heat generated within the antenna may be large if large
magnetic field gradients or rapid gradient switching are
used in the MRI procedure. Secondly, since the signal
from the device is superimposed over the original MR
image, any tissue movement may result in loss in
accuracy of spatial information regarding device
location.

The term "passive visualisation" is used to
describe the case where device visualisation relies on a
difference in magnetic susceptibility between the
material from which the device is made and the
surrounding biological tissue or fluid. In order to
increase this susceptibility difference, it has been
conventional to mark the device with a magnetic (i.e.
paramagnetic, ferrimagnetic, ferromagnetic or
superparamagnetic) material, such as for example
dysprosium oxide ($Dy_2O_3$). To avoid extensive marking
resulting in blurring artefacts in the MR images, it has
been common to incorporate small bands or rings of $Dy_2O_3$
or of a material containing for example 10% w/w $Dy_2O_3$.
*In vitro* studies have shown that such $Dy_2O_3$-marked
devices can be visualised accurately in "near real time"
imaging procedures without causing significant image

artefacts (such as for example blurring or distortion)
as long as the device is parallel to the primary
magnetic field of the MR imager.  Where the device is
perpendicular to the primary field however, blurring and
distortion artefacts occur which result in over-
estimation of device size and decrease in spatial
resolution.

     Biopsy needles are currently monitored using
passive visualisation; however the needles currently
used produce severe image artefacts which distort the
observed size and location of the needles.  To
compensate for this, two techniques have been developed.
Firstly a laser guidance system may be used to guide the
needle to the target tissue; however, using this
technique, it is still difficult to determine needle
depth within the tissue.  Accordingly, needles have been
produced which have a small inner bore at the needle
tip.  This is filled with a gadolinium chelate solution
(e.g. a solution of Gd DTPA, Gd DTPA-bismethylamide or
Gd HP-DO3A) which allows for visualisation of the needle
tip in the MR image.  However the gadolinium chelates
used distribute rapidly into the extracellular space and
thus even with single or multiple injections of the
gadolinium chelate solution only temporary assistance in
passive visualisation of the needle tip is achieved.

     The present invention is directed to facilitating
passive visualisation of invasive devices and relies
instead on introducing into the vasculature a blood pool
MR agent, i.e. an MR contrast agent which does not
distribute into the extracellular space but instead
remains substantially in the intravascular space during
the time course of the visualisation procedure.  The
effect of the contrast agent is to enhance the
relaxation properties of the blood (i.e. to reduce $T_1$
and/or $T_2^*$ for blood) relative to those of the invasive
device.  Thus, using the invention, the invasive devices
traditionally utilised in intraoperative and

interventional MRI can be used.

Thus viewed from one aspect the invention provides a method of interventional or intraoperative MRI wherein an invasive device is inserted into the vasculature of a human or non human animal (e.g. mammalian, avian or reptilian) body or through vascularised tissue in said body and an MR image of at least a part of said body containing said device is generated, the improvement comprising administering into the vasculature of said body by either direct injection of the contrast agent through said device or by i.v. injection of the contrast agent directly into the patient whereby to facilitate visualisation of said device in said image.

Viewed from a further aspect the invention provides the use of a blood pool MR contrast agent for the manufacture of a parenterally administrable MR contrast medium for use in a method of surgery or therapy wherein an invasive device is inserted into the vasculature of a human or non human animal (e.g. mammalian, avian or reptilian) body or through vascularised tissue in said body and an MR image of at least a part of said body containing said device is generated, said method also comprising administering said contrast medium into the vasculature of said body whereby to facilitate visualisation of said device in said image.

By a blood pool MR contrast agent is meant a magnetic (e.g. paramagnetic, ferromagnetic, ferrimagnetic or superparamagnetic) material capable of reducing the $T_1$ and/or $T_2$* of water protons and which if administered into the vascular space does not significantly leak out into the interstitium during the time course of the interventional or intraoperative procedure, i.e. it is essentially confined to the vascular space until excreted or metabolized. Examples of such blood pool agents include polymeric chelates (e.g. cascade polymers or dendrimers carrying metallated chelate groups) and particulates, in particular iron

oxides and liposomes.  Generally the agent should have a
blood half life of at least 5 minutes, preferably at
least 30 minutes.  By way of contrast, the first
parenteral MR contrast agents Gd DTPA (Magnevist® from
Schering), Gd DTPA-bismethylamide (Omniscan® from
Nycomed Amersham) and Gd HP-DO3A (ProHance®) are all
extracellular fluid MR agents; they are water-soluble
mono-chelates which following administration into the
vasculature rapidly extravasate into the interstitium.

Blood pool agents of particular use in the method
of this invention include low molecular weight chelates
which bind to blood proteins, e.g. blood proteins such
as albumin, for example DTPA or DOTA derivatised with
protein binding groups, e.g. lipophilic side chains such
as aromatic moieties, e.g. one or more phenyl ring
systems.  One such example is MS-325/Angiomark of EPIX.

Suitable polymer based contrast agents for use in
the method of the present invention can be carbohydrate
or protein based, e.g. CMD-DTPA-Gd of Guerbet
(Carboxymethyl dextran-GdDTPA conjugates), GdDTPA
polylysine conjugates, or cascade or dendrimer polymers,
e.g. Gadomer 17 of Schering AG or similar cascade
polymers as described in US-A-5874061 (of Schering AG),
herein incorporated by reference.

Suitable iron oxide (or doped iron oxide) based
contrast agents for use in the method of the present
invention are known in the field under the name of SPIO
(superparamagnetic iron oxides) or USPIO (ultrasmall
superparamagnetic iron oxides).  Examples include
carbohydrate stabilised iron oxide particles, e.g.
dextran-stabilised particles such as Combidex of
Advanced Magnetics, and NC100150 (Clariscan, Nycomed
Amersham).

More particularly the magnetic iron oxide contrast
agent is preferably a water-dispersible material
comprising magnetic iron oxide particles having on their
surfaces (e.g. as a coating), an optionally modified

carbohydrate or polysaccharide or derivative thereof,
e.g. a glucose unit containing optionally modified
polysaccharide or derivative thereof, preferably an
optionally modified dextran or starch or derivative
thereof, for example a cleaved (e.g. oxidatively
cleaved) starch or carboxylated dextran. Such iron
oxide complexes preferably also comprise a further
material (e.g. coating material), especially one which
inhibits opsonization, e.g. a hydrophilic polymer,
preferably a functionalized polyalkylene oxide, more
preferably a functionalized polyethylene glycol (PEG),
in particular methoxy PEG phosphate (MPP).

The iron oxide complexes preferably have a core
(i.e. iron oxide particle) diameter (mode diameter) of 1
to 15 nm, more preferably 2-10 nm, especially 3-7 nm, a
total diameter (mode particle size) of 1 to 100 nm, more
preferably 5-50 nm, especially preferably 10-25 nm, an
$r_2/r_1$ ratio at 0.47T and 40°C of less than 3, more
preferably less than 2.3, still more preferably less
than 2.0, especially preferably less than 1.8. The
saturation magnetization (Msat) at 1T is preferably 10
to 100 emu/gFe, more preferably 30-90 emu/gFe.

Other particulate based systems of use in the
method of the present invention include liposomal or
emulsion based agents.

Furthermore, compound 7228 of Advanced Magnetics
can be used in the method of the present invention, as
can the materials described in WO 91/12025, WO 90/01899,
WO 88/00060, WO 91/12526 and WO 95/05669, all to
Advanced Magnetics, and those described in WO92/11037
and WO90/01295, all of which publications are
incorporated herein by reference.

Using blood pool MR contrast agents which remain in
the vascular space during the MRI-guided procedure
allows for extensive monitoring of the invasive device
during the procedure. The method of the invention, an
"induced passive visualisation" technique, not only

allows visualisation of static devices, such as stents,
but also can be used to guide placement of devices, e.g.
it can be used in ablation therapy to mark vessels
and/or aid in the placement of the ablation device.

Currently, three types of ablation procedures are
in use clinically: interstitial laser-induced
thermotherapy (LITT); focussed ultrasound; and rf-
ablation.

LITT is used for the destruction of local tumors in
solid organs - laser light is delivered to the tumor
through optical fibres and tumor destruction occurs by
direct heating by the laser light. The optical fibre is
introduced in MRI-guided procedures using an MRI-guided
catheter; induced passive visualisation according to the
invention will allow accurate visualisation of the
catheter and ensure correct placement of the optical
fibre tip.

During MRI-guided focussed ultrasound ablation
therapy, an ultrasound transducer is moved hydraulically
across the patient and the depth and position of the
ultrasound focus are determined by laser optical fibres
which are locked onto target from an MR image. Induced
passive visualisation according to the invention will
assist by increasing the accuracy of the focus
positioning since the technique allows for monitoring of
all major vessels surrounding the tumor, as well as of
the tumor itself, throughout the procedure.

In rf-ablation therapy, a radiofrequency electrode
is placed in tissue which has previously been injected
with saline. The tissue around the electrode is then
heated by applying 1500-1600 mA to the electrode. The
technique is relatively non-specific since treatment
regions vary in shape and are dependent on local saline
concentration; induced passive visualisation according
to the invention will increase specificity by allowing
more accurate placement of the electrode and more
accurate estimation of saline concentration. Again,

both tumor and surrounding vessels may be monitored
throughout the procedure.

During all intraoperative and interventional
procedures, complications due to bleeding increase the
risks associated with the procedures. Induced passive
visualisation according to the invention however allows
for accurate monitoring and assessment of bleeding
during the procedure. Since the blood pool MR contrast
agent remains in the blood during the procedure, any
damage to blood vessels that causes changes in vessel
permeability can be observed.

The invasive devices which can be monitored
according to the invention include, but are not limited
to, catheters, balloons, optical fibres, guide wires,
needles (e.g. biopsy needles), electrodes, electrode
leads, implants, stents and stent graphs. Generally
these devices will be diamagnetic and exhibit long $T_1$
values. If desired, the devices may be marked with a
magnetic susceptibility agent, e.g. bands or strips
containing dysprosium oxide - however such marking is
not necessary and desirably is avoided. As in all MRI-
guided procedures however the devices used are
preferably not substantially ferromagnetic or
ferrimagnetic as this will cause image defects and
gradient switching may cause unwanted motion of the
devices.

The MR imaging procedure used in the method of the
invention may for example be any conventional MRI
procedure, e.g. $T_1$ or $T_2$* weighted spin echo or gradient
echo sequences. However, fast imaging procedures, such
as gradient echo and echo planar imaging procedures are
preferred. In a particularly preferred embodiment of
the method of the invention, the imaging procedure
involves administration of an iron oxide blood pool MR
contrast agent, gradient echo imaging using small flip
angles (e.g. 10 to 45°) and short echo times (e.g. 0.5
to 5 ms) and using larger flip angles (e.g 55 to 75°)

and longer echo times (e.g. 6 to 20 ms). In the larger
flip angle/longer echo time images the signal from the
contrast agent containing blood is diminished and, using
both images, visualisation of the invasive device may be
facilitated (especially where the device contains a
gadolinium chelate solution as a marker).

Viewed from another aspect the invention provides a
method of interventional or intraoperative MRI wherein
an invasive device is inserted into the vasculature of a
human or non human animal body or through vascularised
tissue in said body and an MR image of at least a part
of said body containing said device is generated, the
improvement comprising administering into said body a
contrast agent which remains in the vasculature during
the time course of said method so that the difference in
$T_1$ and/or $T_2$ and/or $T_2^*$ between the blood and said device
containing a paramagnetic or diamagnetic material may be
utilized to generate image contrast between blood and
said device.

By containing it is meant that the device may
contain the paramagnetic or diagnostic material when it
is inserted or that the material may be placed in the
device following its insertion.

Thus the method may involve administering a blood
pool contrast agent (e.g. an USPIO) into the vasculature
and filling an invasive device (before or after its
insertion) with a paramagnetic agent (e.g. a gadolinium
chelate) and then generating an image of the vasculature
using very fast, heavily $T_1$-weighted sequences. In this
way slice position matching with the device is
unnecessary. The selectivity of differentiation between
blood and device is achieved by a minor increase in echo
time that would not normally result in the sequence
being termed non-$T_1$-weighted.

Viewed from a further aspect the invention also
provides methods of interventional or intraoperative MRI
wherein an invasive device is inserted into the

vasculature of a human or non human animal (e.g.
mammalian, avian or reptilian) body or through
vascularised tissue in said body and an MR image of at
least a part of said body containing said device is
generated, the improvement comprising administering into
said body a contrast agent which remains in the
vasculature during the time course of said method so
that the $T_1$ and/or $T_2$* of the blood is enhanced relative
to that of said device, and wherein when enhancing the $T_1$
of the blood relative to said device, $T_1$-weighted
sequences should be used and said device should be
filled with diamagnetic material (e.g. saline or
medication) so that the blood appears bright relative to
said device, and wherein when enhancing the $T_2$* of the
blood relative to the device, $T_2$ or $T_2$*-weighted
sequences should be used and the device coated with or
filled with paramagnetic material (e.g. Gd complex or Mn
complex) so that said device appears bright relative to
the blood.

By initially or subsequently containing it is meant
that the device may contain the paramagnetic chelate
(preferably a water-soluble Gd or Mn monochelate) when
it is inserted or that the chelate may be filled into
the device following insertion.

In both the methods of the invention it is
desirable to use as the blood pool MR contrast agent a
superparamagnetic iron oxide, optionally coated with an
opsonization inhibitor such as PEG, e.g. as described in
WO97/25073. Such particles allow $T_1$ effects to
predominate in heavily $T_1$-weighted imaging sequences -
leading to an increased signal intensity for blood
relative to invasive device (i.e. a "bright blood"
technique) - and allow $T_2$* effects to predominate when
$T_2$-weighted imaging sequences are used - leading to a
decreased signal intensity for blood relative to
invasive device (i.e. a "black blood" technique). Such
black blood techniques are used where paramagnetic

devices or devices with paramagnetic markers are used.
The bright blood technique is used with diamagnetic
devices which contain a diamagnetic material (e.g.
saline or medication).

The dosage of the contrast agent used according to
the invention will depend upon the species, the
longitudinal relaxivity of the agent, the magnetic
moment of the agent at the imaging field strength and
the sequence parameters used to acquire the image.
Desirably the blood pool MR contrast agent is
administered at dosages sufficient to achieve $T_1$ values
in blood, at steady state, of less than 300 ms, more
preferably less than 200 ms and still more preferably
less than 100 ms.

In MR procedures in which a blood pool contrast
agent is to be administered into the vasculature, this
would normally be done outside the region of interest
for imaging, e.g. in a peripheral vein and one would not
expect images of the catheter to be generated either on
contrast agent administration or on contrast agent
circulation (or recirculation) past the catheter. Thus
in the methods of the invention the invasive device may
be a catheter used for administration of the blood pool
contrast agent; however more generally and preferably
the invasive device will be other than a device through
which the blood pool contrast agent is administered.

The blood pool contrast agent will preferably be
administered by injection or infusion into the
vasculature, for example infused over periods of 2
seconds to 5 minutes.

The contrast agent will desirably be formulated in
a sterile aqueous medium, optionally containing further
excipients such as pH modifiers, osmolality modifiers,
chelating agents, etc.

The methods of the invention will now be described
further with reference to the following non-limiting
Examples and the accompanying drawings, in which:

Figure 1 shows the construction of two phantoms use to study the feasibility of catheter tracking by MRI;

Figure 2 shows a bright blood image of phantoms having catheters filled with saline (top) or 10mM Omniscan (bottom) and placed perpendicular to the applied field;

Figure 3 shows a bright blood image of phantoms having catheters filled with saline (left) or 10mM Omniscan (right) and placed parallel to the applied field;

Figure 4 shows a dark blood image of a phantom having a catheter filled with 10 mM Omniscan and placed perpendicular to the applied field;

Figure 5 shows a summary of an MRI signal optimisation study conducted at TR/TE=5.0/1.5 ms (Fig 5a) and TR/TE=18.0/9.0 ms (Fig 5b) with varying flip angles and concentrations;

Figure 6 shows spoiled GRE images of phantoms collected with parameters TR/TE/FA=5.2/1.2/40° (Fig 6a) and TR/TE/FA=18/9.0/70° (Fig 6b); and;

Figure 7 shows spoiled GRE *in vivo* images of the abdomen of a pig collected with parameters TR/TE/FA=5.2/1.2/40° (Fig 7a) and TR/TE/FA=18/9.0/70° (Fig 7b).

An aqueous suspension superparamagnetic iron oxide MR contrast agent, prepared according to the description of Example 12 of WO 97/25073 was used in these Examples.

The characteristics of the suspension were as follows:

[Fe] = 30.2 mg Fe/ml, density = 1.0589 g/ml, $r_1$ = 19.3 $s^{-1}mM^{-1}$, $r_2$ = 31.2 $s^{-1}mM^{-1}$, $r_2/r_1$ = 1.61, saturation magnetisation (Msat) = 84 emu/g Fe.

## Example 1

## Phantom Study

A simple *ex vivo* phantom study was performed in order to assess the feasibility of performing induced passive catheter tracking by reducing $T_1$ and $T_2^*$ of blood. Two phantoms were prepared by placing sections of a conventional interventional catheter (Pebax Souple 5F with inner diameter of 1.17 mm and outer diameter of 1.69 mm) into two plastic tubes of 10 mm diameter and 85 mm length inside 21 mm diameter, 60 mm length glass vials. The plastic tubes were fixed in the vial in a 2% agar gel containing gadolinium polymers (e.g. a gadolinium polychelate such as polylysinepolyDTPA.Gd) so that the $T_1$ of the gel at 40°C was 535 ms. The plastic tubes were filled with fresh human blood (Hct = 47%) containing sodium heparin. The iron oxide contrast medium was added to the blood so that the concentration of Fe added was 1.0 mM Fe (which is equivalent to a dose of 4 mg Fe/Kg bodyweight). One of the catheters was then filled with saline and sealed and the other catheter was filled with 10 mM Omniscan® (10 mM Gd DTPA-bismethylamide) and sealed. The arrangement was as shown in Figure 1 of the accompanying drawings. All imaging was performed at 1.5 T (Philips Gyroscan ACS-NT) using $T_1$ and $T_2$-weighted 3D-gradient echo sequences (FFE). Table 1 shows the sequence parameters which were kept constant for all experiments.

**Table 1:** Pulse Sequence parameters for static *in vivo* phantom imaging

| Parameter | Value |
|-----------|-------|
| TR | 15.4 ms |
| Slice thickness | 0.7 mm |
| Field of view | 140 x 140 |
| Nmat | 256 x 256 |
| NEX | 2 |
| Flip angle | 30° |

The following two imaging series were performed on the phantoms:

Series 1: Phantoms placed perpendicular relative to the applied field. Images were acquired at echo times (TE) of 2 ms (for bright blood) and 12 ms (for black blood).

The phantoms were removed from the magnet and inverted several times to ensure the homogeneity of the blood samples prior to the on set of imaging Series 2.

Series 2: Phantoms placed parallel relative to the applied magnetic field. Images were acquired using a TE of 2 ms so that bright blood was obtained.

Figures 2, 3 and 4 of the accompanying drawings show the results obtained when either the $T_1$ (bright blood obtained using short echo times) or $T_2^*$ (dark blood obtained using long echo times) of the blood is reduced relative to the device. In Figures 2 and 4 the phantoms were placed perpendicular relative to the applied magnetic field and Figure 3 shows the phantom placed parallel relative to the applied field. In Figures 2 and 3, short echo times were used so that the $T_1$ of the blood was reduced relative to that of the catheters. Figure 2 shows the phantoms placed perpendicular to the applied field. The catheters were filled with either

- 15 -

saline (top) and 10 mM Gd DTPA-BMA (bottom). In Figure
3, the phantoms were placed parallel to the applied
magnetic field and the catheter containing the saline is
shown to the left and the catheter containing the Gd
DTPA-BMA is shown to the right. The diameter shown in
these figures represents the total catheter diameter
estimated by MRI. The actual catheter diameter for all
phantoms was 1.69 mm.

The catheter filled with saline represents a
conventional interventional catheter. The catheter
filled with 10 mM Omniscan® represents an MRI catheter
containing a paramagnetic tracer. The results clearly
indicate the size of the catheter (total diameter) was
underestimated when the catheter was filled with 10 mM
Omniscan®. However, when the catheters were filled with
saline (which represents a normal interventional
procedure) fairly accurate estimates of size were
obtained. The under-estimation of size observed when
using a paramagnetic tracer is most likely due to the
fact that the wall of the catheter has very few protons
and consequently is not MR-visible. This means that the
signal coming from the outer wall of the catheter is
dark relative to the signal inside the catheter (which
contains 10 mM Gd) and signal of the blood (which
contains the iron oxide contrast medium). As a result,
the size reflects only the inner-diameter of the
catheter. However, when saline is used, a homogeneous
dark signal is observed which accurately reflects
catheter size. This is due to the fact that the
catheter and the saline filling the catheter both appear
hypointense relative to the blood containing the iron
oxide contrast medium (bright blood obtained using short
echo times).

No artefacts were observed for either phantom in
any orientation relative to the applied field.

Figure 4 shows the results obtained when the $T_2*$ of
the blood is reduced relative to that of the device by

- 16 -

increasing the echo times used. Here the blood appears
black since the reduction of $T_2$* dominates the observed
signal. For "black blood", only the catheter filled
with 10 mM Gd (representing the paramagnetic tracer
catheter) is visible. The diameter shown represents the
diameter estimated by MRI. The actual diameter was 1.69
mm. The phantom was placed perpendicular relative to
the main field. The size of the catheter obtained by MR
still reflects only the inner-diameter.


## Example 2

Example 2 summarises a study performed in order to
determine the optimal imaging sequence parameters
required for passive induced visualisation when black
blood techniques are employed (catheter filled with a
paramagnetic tracer, blood containing the iron oxide
contrast medium and long echo times used to enhance the
$T_2$* of the blood relative to the catheter).

Figures 5A (TR/TE=5.0/1.5 ms) and 5B
(TR/TE=18.0/9.0 ms) show experimental signal-intensity
profiles of diluted Gd solutions as a function of
concentration and flip angle. Brighter zones correspond
to higher signal intensity. Note the shift of the
signal maximum towards lower concentrations for longer
echo times.

A catheter (0.8 mm inner diameter) with an
Omniscan®/$H_2O$ (0.01 M) filled guide-wire lumen was
inserted into a flexible tube (i.d. 5 mm) filled with
the iron oxide contrast medium (0.5 mg Fe/ml). Spoiled
2D and 3D GRE images (2D imaging without slice
selection) were collected with two different echo times
(NEX=1, 256x192x20, FOV=28x14 cm) as shown in Figures 6A
(3D TR/TE/FA=5.2/1.2/40°) and 6B (2D TR/TE/FA=18/9/70°).


## Example 3

In this example the black blood technique was used

*in vivo* by accessing the right femoral artery of two pigs under general anaesthesia. For display of the vascular system the iron oxide contrast medium was administered intravenously at a dose of 5 mg Fe/kg bodyweight. Via the femoral approach, a 5F PTA catheter was introduced into the abdominal aorta. The 40x12 mm balloon was filled with a 7.7 mM Gd solution of GdDTPA-bismethyl-amide and imaged with spoiled GRE sequences. The experiment was repeated with a 6F catheter in which the guide-wire lumen was filled with the Gd-solution. Figures 7A (3D TR/TE/FA=5.2/1.2/40°) and 7B (2D TR/TE/FA=18/9/70°) show the resulting spoiled GRE *in vivo* images of the abdomen of the pig. Note the bright signal intensity of both vascular system and catheter balloon in the short-echo image. With TE=9 ms only the balloon (arrow) and blood in the intraperitoneal cavity from a preceding experiment remain visible.